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Fatigue of dental ceramics

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FATIGUE OF DENTAL CERAMICS

ABSTRACT

Objectives. Clinical data on survival rates reveal that all-ceramic dental prostheses are susceptible to fracture from repetitive occlusal loading. The objective of this review is to examine the underlying mechanisms of fatigue in current and future dental ceramics.

Data/sources. The nature of various fatigue modes is elucidated using fracture test data on ceramic layer specimens from the dental and biomechanics literature.

Conclusions. Failure modes can change over a lifetime, depending on restoration geometry, loading conditions and material properties. Modes that operate in single-cycle loading may be dominated by alternative modes in multi-cycle loading. While *post-mortem* examination of failed prostheses can determine the sources of certain fractures, the evolution of these fractures *en route* to failure remains poorly understood. Whereas it is commonly held that loss of load-bearing capacity of dental ceramics in repetitive loading is attributable to chemically-assisted 'slow crack growth' in the presence of water, we demonstrate the existence of more deleterious fatigue mechanisms, mechanical rather than chemical in nature. Neglecting to account for mechanical fatigue can lead to gross overestimates in predicted survival rates. *Clinical significance.* Strategies for prolonging the clinical lifetimes of ceramic restorations are proposed based on a crack-containment philosophy.

KEY WORDS: dental ceramics, fatigue, fracture modes, lifetime, crowns, fixed-partial dentures

1. Introduction

Major dental restorations such as crowns and fixed-partial dentures (FDPs), as well as other biomechanical prostheses, are experiencing a rapid shift toward ceramic materials, partially for their strength and bioinertness but more so for their aesthetics.¹⁻³ However, ceramics are brittle and susceptible to fatigue fracture in repetitive function. Although occlusal loading is nominally compressive, with bite forces supported in individual 'dome-like' structures (crowns) or in frameworks with connectors (FDPs), some tensile stresses are inevitable. Cracks tend to follow paths where these tensile stresses are greatest. While a ceramic restoration may fracture abruptly from a single intense overload, it is more likely that failure will occur cumulatively after an extended period of seemingly innocuous but lower-load biting events. Such fractures are manifest in the clinical literature as 'lifetime' or 'survival rate' data. Beyond such data lies a burning question: what are the underlying physical bases for designing next-generation ceramic materials for greater long-term performance?

The drive toward ceramic restorations is fraught with compromise.^{2, 4} There is a perception that ceramic crowns and FDPs are not yet as reliable as those with traditional metal-frameworks.¹ The ceramics with the most desirable aesthetics, notably porcelains, tend also to have the lowest resistance to crack propagation ('toughness').⁵⁻⁷ Conversely, tougher ceramics such as aluminas and zirconias^{5, 8} are not generally aesthetic. Glass-ceramics^{9, 10} occupy a middle ground. Two well-grounded routes exist to overcome these countervailing tendencies. The first is to bond an aesthetic porcelain veneer onto a stiff alumina or zirconia core to provide support in flexural loading.^{1, 2, 11} However, the veneer remains a weak link, susceptible to chipping and delamination from the core (although as will be demonstrated later the core itself is not immune). Coefficient of thermal expansion (CTE) mismatch between veneer and core and low thermal diffusivities in most ceramics can lead to deleterious tensile stresses within the bilayer during heat treatment.¹²⁻¹⁶ The second route is to develop crack-resistant but partially translucent monolith ceramics to circumvent the need for veneering altogether—e.g. lithium disilicate glass-ceramics (IPS e.max Press or CAD by Ivoclar-Vivadent),¹⁷ or zirconias with fine grains (e.g. Lava Plus by 3M ESPE,

Bruxzir by Glidewell, Allzir by New Image) or surface-infiltrated glass.¹⁸⁻²² Monolith ceramics also avoid weak veneer/core interfaces, minimizing the risk of delamination. In both routes, zirconia-based ceramics are emerging as materials of choice.

Given the brittleness of ceramics, it is hardly surprising that prosthetic failures do occur. Some of the more commonly observed clinical fracture modes are sketched in Fig. 1. They include cracks initiating from the contact zone at the occlusal surface,²³ from the cementation surface beneath the contact,²⁴ and from the margins of crowns and connectors in FDPs.²⁵⁻³¹ Some examples of clinically fractured prostheses are shown in Figs. 2a to 2c, revealing fracture from a wear facet on a porcelain-veneered zirconia crown occlusal surface, a longitudinal crack initiated from the margin of a Dicor glass–ceramic crown, and a flexure crack at the connectors of a porcelain-veneered zirconia FDP. All of these cracks can result in severe damage or irrecoverable failure. Chipping fractures initiate from contact damage sites and detach at least part of the veneer from the core. Through-thickness fractures initiate from the occlusal or cementation surface beneath the contact or from the margins or connectors and can split a prosthesis in two. Clinical trials reporting survival rates for several all-ceramic systems indicate vulnerabilities to all these fractures.^{25-27, 32-48} Broadly speaking, porcelain-veneered systems show higher fracture rates than full-contour monoliths, FDPs more than single crowns, and glass–ceramic more than zirconia monoliths, although the variability in data from study to study can be high.

The physical mechanisms of fatigue in ceramic restorative materials have not been well documented in the dental literature. The prevailing view, borrowed originally from fundamental studies in the materials science community,^{49, 50} is that fatigue can be accounted for by chemically-enhanced, rate-dependent crack growth in the presence of moisture.⁵¹⁻⁶⁰ According to this viewpoint, water enters incipient fissures and breaks down cohesive bonds holding the crack walls together.^{49, 61} The result is so-called 'subcritical' or 'slow' crack growth (SCG) which progresses steadily over time, accelerating at higher stress levels and ultimately leading to failure. The notion is attractive because it lends itself to rigorous 'fracture mechanics' analysis in terms of

explicit crack velocity equations, enabling one to predict lifetimes in terms of specified stress states.⁶² But recent studies in the materials science arena reveal that fatigue is more complex than just SCG. In addition to *chemical* degradation, there are mechanisms of *mechanical* degradation that can augment the fatigue process.^{12, 63-69} Mechanical fatigue operates exclusively in cyclic loading and cannot be inferred from static or monotonic loading tests. It can be relatively destructive, meaning that predictions based exclusively on SCG assumptions may grossly overestimate potential lifetimes. 'Fractography'⁷⁰—the microscopic analysis of post-failure restorations—can point to likely starting sources of fracture but is limited in its capacity to shed light on the fatigue mechanisms themselves, or to determine the sometimes complex evolutionary progression of competing fractures to completion.

It is important to understand the interplay between competing fracture modes in order that the best fatigue-resistant restorative ceramics may be developed. Accordingly, this article surveys the fatigue behavior of commonly used dental ceramics from a biomechanics point of view. The principal mechanisms by which chemical and mechanical fatigue occur are outlined. Simulated occlusal loading tests on model flat layer specimens (as well as on anatomically-correct prostheses), designed to represent essential features of dental ceramic layer restorations bonded to a relatively compliant dentin substrate, enable various competing fracture modes to be identified and quantified in a clinically relevant context. Strategies for prolonging the life of ceramic restorations are explored.

2. Failure Evaluation

2.1 Fracture modes

Failures in dental ceramic prostheses are usually associated with structural defects or 'flaws'. Flaws may arise during fabrication and preparation, or from post-placement chewing activity.⁷¹ They can take the form of microstructural defects within the ceramic, from machining in fabrication or sandblast damage during fitting,^{69, 72, 73} from wear facets and contact damage on the occlusal surface⁷⁴ or cementation⁶⁹ surfaces, or from micro-contacts with hard sharp objects.⁶⁷ In

ceramics, flaws generally assume the form of microcracks of sub-millimeter scale, often below visual detection. Valuable clues as to the origin of such flaws can be provided from post-failure fractography.⁷⁰ It follows that good fabrication procedures and avoidance of preparation surface damage may be crucial elements of prosthetic dentistry. But this linking of fracture with flaw populations is to belie the essence of the failure process. Most often, newly formed cracks are 'contained'—they first arrest and subsequently extend incrementally over a long cycling period prior to ultimate failure. In natural teeth this crack 'stability' is manifest as closed fissures or 'lamellae' along the enamel walls.⁷⁵⁻⁷⁹ It is conceivable that steady crack growth could be monitored by periodic inspections of prostheses *in vivo*, but this is beyond the scope of normal dental practice, and in any case there is no guarantee that critical damage will be visible at the outer surface of a near-opaque restoration. Moreover, different modes of fracture can dominate under certain geometric conditions, and at different stages in the loading. Consequently, fracture evolution is complex and difficult to infer from conventional *post-mortem* and *in vivo* examinations alone.

What is missing from clinical studies is a fundamental understanding of the various mechanisms by which flaws evolve into full-scale fractures, especially in long-term cyclic loading. One approach is to conduct laboratory tests on anatomically-correct specimens by pressing down directly at an exposed surface with an indenting plate or sphere. Examples of cracked porcelain-veneered zirconia prostheses are included in Fig. 2, for crowns loaded vertically at the edge of a buccal cusp (Fig. 2d)⁸⁰, at the lingual aspect of a buccal cusp with sliding motion toward the central fossa (Fig. 2e),⁸¹ and for a 3-unit FDP loaded at the buccal cusp of the pontic (Fig. 2f).⁸² However, such complex structures are not amenable to simple analysis and prediction. It is accordingly expedient to conduct *ex vivo* tests on model brittle specimens that retain the essential material and geometrical features of crowns, but in an idealized way that enables *in situ* monitoring of individual or concurrent cracks from initiation to full failure. In this approach, tests are carried out on flat-layer plate⁸³⁻⁹⁰ and dome-like shell⁹¹⁻⁹⁴ structures bonded to a polymeric substrate representative of a compliant dentin-like support. The undersurfaces of the

test specimens can be given different preparations representative of clinical protocols, to examine the effect of surface finish. The specimens are top-surface-loaded with a spherical indenter, representative of occlusal contact. The arrangement allows for variations in contact conditions—single-cycle axial, off-axis⁹³ and sliding,^{81, 95, 96} or cyclic.^{12, 65, 66, 74, 97-99} Generally, cracking begins at either the top occlusal or intaglio cementation surface, or sometimes, in the case of shell structures, at the margins. Damage from tests on opaque plate or shell specimens can be examined by sectioning techniques,^{100, 101} but such tests are data-limited and labor-intensive. Simpler and more informative are tests on systems constructed from all-transparent materials as proxies for porcelain veneer and hard ceramic monolith or core, enabling video monitoring of fracture during an actual testing cycle. The contacting sphere can be made of hard material to represent biting on a hard object, or polymer to represent chewing on soft food.¹⁰² While it is acknowledged at the outset that flat and shell model systems of this kind neglect certain important fine details, e.g. convoluted cuspal geometry and wall thickness variations, they nevertheless provide a powerful physical basis for understanding and analyzing how clinical restorative prostheses fail.

Images of top-surface contact cracks in porcelain are shown in Fig. 3, for tests in aqueous environment. The examples include half-surface and side views of sectioned flat-surface specimens in multi-cycle axial (Fig. 3a) and tangential loading (Fig. 3b) with a hard sphere. The damage is precipitated by the inordinately high local stress concentrations around the contact,¹⁰³ and is most common with low-radius spheres. The contact 'footprint' is akin to a wear facet, with near-surface microplastic and microcrack damage, which can act as a precursor to occlusal cracks, of which there are several variants.^{5, 101, 103} In normal loading (Fig. 3a), 'outer' and 'inner' axisymmetric 'cone' cracks (O and I cracks in Fig. 1a) initiate just outside and within the contact circle and extend deep into the subsurface. The former can occur in a single heavy cycle, and thereafter grow steadily with time under load by SCG. The latter appear only after prolonged multi-cycling, and are driven mechanically by hydraulic pumping of fluid into surface microcracks.⁶⁸ In sliding loading (Fig. 3b), a tangential component skews the tensile stress field,

with attendant development of asymmetric, partial cone cracks (P cracks, Fig. 1a) at the trailing edge of the contact.^{96, 104-107} Like inner cones, partial cones grow more rapidly in multi-cycle loading, again suggestive of some hydraulic pumping. With continued cycling at sufficiently high load, cone cracks can penetrate through a veneer layer to the core interface, or even through a monolith layer to the cementation surface, with consequent delamination. In addition, 'median' cracks (M cracks, Fig. 1a) on planes containing the load axis may extend downward from more severe surface damage zones,^{66, 108, 109} from contacts with small spheres or sharp points. Edge chipping (C cracks, Fig. 1a) can be considered a special case of median or cone cracking in the vicinity of a top-surface edge.¹¹⁰ Analogous tests on model structures reveal that chipping fracture is not abrupt, but that the crack extends steadily downward with increasing load (or number of cycles) prior to instability.¹¹⁰

Fractures initiated away from the top-surface contact zone are shown in Fig. 4, for model transparent layer systems bonded to a dentin-like resin base. Figures 4a and 4b show side views of subsurface radial cracks (R cracks, Fig. 1a), rendered visible by interfacial interference fringes, in a glass monolith layer and glass/sapphire bilayer under load, for tests in surface loading with a hard sphere. The tensile stresses responsible for these fractures are 'flexural' in nature,⁸⁵ and are much lower in magnitude than contact stresses but also much less concentrated. The radial cracks have initiated at the cementation surface beneath the contact and spread sideways and upward. Viewed from below, they are 'star-shaped' with multiple arms.^{87, 111} They tend to close up during unloading, causing the interference fringe patterns to disappear and thus make detection difficult. The same cracks are even harder to detect in opaque or translucent ceramics until they break through to the top surface, or until delamination occurs at either the cementation or veneer/core interface. The stress state in the shells is a little more complex.^{91, 112} Figures 4c and 4d show radial cracks in resin-filled glass shell structures loaded with a hard and soft sphere, respectively.^{91, 113} A hard contact initiates the same kind of radial cracks seen in Figs. 4a and 4b. A soft contact engulfs the top surface within a compression zone, suppressing radial cracking there and transferring tensile stresses to the margin⁹⁴—the result is the same kind of longitudinal

fracture, at similar failure loads, but with the cracks propagating in an opposite direction.

In all cases in Figs. 3 and 4 the fractures remain contained within the brittle outerlayer. Once radial cracks break through the shell thickness they progress slowly but inexorably with continued cycling around the side walls, under the influence of SCG.^{114, 115} The ensuing full fractures have all the essential characteristics of the clinical failures depicted in Fig. 1. Even contacts that produce no evidence of surface damage in a single load cycle can lead to catastrophic crack growth over time. Severe overloads can lead to delamination or even penetration of the cracks into the dentin-like sublayer.⁹¹

2.2 *Strength data*

Fatigue evaluation of individual dental ceramics has been conducted using standard flexure testing methodologies. The most common method is to break flat bars or disks, and to evaluate the maximum tensile stress S ('strength') as a function of number of cycles n . The advantage of such testing is simplicity in specimen preparation and data accumulation. A useful variant is first to bond the bar or plate to a dentin-like polymer base and then load the top surface sinusoidally with a hard sphere at a specified frequency until a radial crack abruptly initiates at the cementation surface, as in the arrangement of Fig. 4a. Such a supported layer structure is one step closer to the clinical reality of crown/dentin configurations. A video camera placed beneath the bilayer specimens enables detection of radial crack initiation, even in opaque materials.¹¹¹

Strength data obtained in this way are shown in Fig. 5, for fine-grain zirconia plates (Prozyl Y-TZP, Norton, East Granby, CT) of thickness 0.6 mm loaded at frequency 10 Hz.^{63, 67} The symbols represent intaglio surfaces subjected to different treatments: high polish (1 μm diamond paste), sandblasted (50 μm alumina particles), and pre-indented with a sharp Vickers diamond at 0.1 and 10 N (approximately 1 μm and 10 μm half-diagonal impressions, i.e. on a scale typical of linear scratching from an errant hard particle). The data points indicate individual tests, the straight lines predicted strength falloffs due exclusively to chemically-assisted crack growth, and the curved lines empirical data fits. Several conclusions may be drawn from Fig. 5: (i) polished

surfaces diminish in strength by about a factor of two or more over the cyclic range (equivalent to 5 years or more at normal biting frequency), consistent with expectation from SCG; (ii) sandblasting degrades the strength of the zirconia, in this case by about a third relative to polished surfaces, indicative of the introduction of microcracks from the particle abrasion, but still consistent with SCG⁶⁹; (iii) contact with individual hard particulates causes a more rapid strength drop, indicative of superposed mechanical degradation.^{69, 97, 116} Comparative strength tests in monotonically sustained loading over equivalent test durations show no such deviations from linear SCG predictions, providing diagnostic confirmation of a mechanical fatigue component.⁶⁷

These trends are representative of all dental ceramics—it is just the vertical positions on the plot that differ. Typically, the strength levels for alumina-based ceramics and lithium disilicate glass-ceramics are about one half to one third those for zirconia, while the levels for porcelain are about an order of magnitude lower.¹¹⁷

2.3 *Lifetime data*

Strength testing tells only part of the fatigue story. Crack initiation at a maximum tensile stress does not necessarily signify 'failure' of a complex clinical layer system. As demonstrated in Figs. 3 and 4, newly formed cracks arrest within the layer interior. Additional cycling, or single-cycle overload, is then required to drive the cracks to full penetration through the layer and outward to the edges or margins. It is in this context that transparent model structures such as those in Fig. 4 provide a powerful means for following all stages of fracture in cyclic loading, culminating in materials databases and predictive fracture mechanics relations that enable estimates of lifetimes for more clinically representative all-ceramic systems.^{12, 65, 69, 118}

To illustrate, Fig. 6 plots through-section crack depth versus number of cycles for flat glass plates of thickness 1 mm bonded to a thicker dentin-like polymer base (cf. Fig. 4a) and loaded with an axial force at its top surface by a hard sphere at frequency 1 Hz.^{65, 66} The surfaces of the glass plate were abraded with SiC grit (analogous to flaws and defects in porcelain) to provide starting flaws for the fractures. Figure 6a plots growth behavior for downward-extending cracks

initiating at the top surface—outer and inner cone cracks and median cracks in axial loading, partial cones in sliding loading. Figure 6b shows behavior for upward-extending radial cracks initiating at the intaglio surface. Vertical portions of some of the lines indicate abrupt initiation at later stages of the cycling. For reference, the line for outer cone cracks in Fig. 6a is entirely consistent with SCG, indicated by the dashed line. Inner cones and median cracks do not initiate until much later in the cyclic history, but then overtake the outer cones with accelerated penetration. In sliding contact, partial cones initiate much earlier, and lead to even more premature failure. The strong upward deviation of the inner and partial cone and median cracks from the trendline for outer cone cracks is indicative of a mechanical component in the fatigue response. The subsurface radial cracks in Fig. 6b also initiate later in the cyclic history, but then extend nearly parallel to the SCG trendline in Fig. 6a.

In all cases in Fig. 6, there is substantial stable crack extension between crack initiation and final layer penetration, meaning that even well-developed cracks can be contained within the structure during the oral history. Such plots usefully demonstrate the interplay between different fracture modes. A mode that dominates in single-cycle loading can be completely overtaken by a competing mode after continued cycling. The absolute and even relative positions of these curves can shift around with changes in biting force, material system, internal residual stress states, and layer thickness, with resultant crossovers in dominant fracture mode.^{65, 69, 118} In actual curved-surface prostheses, post-initiation fracture stability becomes less pronounced with declining smaller tooth size, until ultimately failure may occur spontaneously from a newly initiated crack, i.e. without the stable phase.^{91, 113} The testing methodology is readily extended to trilayers, with the critical crack configuration now defined by intersection with the internal veneer/core interface (e.g. Fig. 4b).¹² Top-surface veneer cracks can then cause delamination at the interface, while bottom-surface core cracks are more likely to penetrate abruptly across the interface into the veneer.¹¹⁹ In the context of prosthetic failures, little of the complex crack history evident in data sets such as those in Fig. 6 is amenable to inference from *in vivo* inspection of outer surfaces or *ex vivo* inspection of remaining parts, and certainly not from any single-cycle tests.

Data such as those in Fig. 6, in combination with fracture mechanics and finite element modeling, facilitate the derivation of explicit relations for critical bite forces for full layer penetration in terms of important clinical material properties (notably elastic modulus and toughness) and geometrical dimensions (contact dimensions, layer thickness, curvature).^{12, 65, 66, 69, 112, 118, 120} These relations, in conjunction with data extrapolations, enable lifetime predictions for any given clinically relevant ceramic layer configuration to be plotted on 'damage maps' as critical number of cycles versus bite force. Examples are shown in Figs. 7a and 7b for flat porcelain-veneered lithium disilicate and zirconia ceramic layers cemented to a dentin base (porcelain thickness 1 mm), and in Figs. 7c and 7d for their lithium disilicate and zirconia monolith counterparts, in each case with net layer thickness 1.5 mm and axial loading with opposing porcelain or enamel surface of radius 5 mm. The linear plots represent occlusal surface cone cracks and cementation subsurface radial cracks, all showing progressive declines in sustainable bite forces with increasing cycling. In the case of the veneered layers, failure is dominated by cone cracks, with a switch from outer to inner at longer cycling times. In the monoliths, no plots are shown for cone cracks, since none form at all at the load range and sphere size represented, meaning that radial cracks comprise the more likely source of fracture. With due acknowledgement of approximations in the analyses and uncertainties in fracture parameters, estimates of critical biting forces over any given number of cycles are probably not more accurate than $\pm 25\%$. Again, the curves in Fig. 7 will shift around with changes in key material and geometric variables (Sect. 4). Notwithstanding these caveats, fracture maps of the kind in Fig. 7 provide valuable quantitative insight into the prospective lifetimes of prescribed material systems.

The data in Fig. 7 correspond to normal loading along a tooth axis with a relatively blunt contact. Loading with a sphere of smaller radius is not likely to shift the radial crack plots substantially, but can exacerbate the onset of microplasticity at the top surface, with consequent increased likelihood of median cracking and accelerated wear (see Sect. 3). Superposition of a sliding component can produce partial cones and biting on a hard contact can induce median cracks (Fig. 6a), each of which can dominate the failure process under certain conditions. Not

specifically represented in Figs. 6 and 7 is edge chipping, from loading with a sharp point close to the side wall of a prosthesis. While no systematic fatigue data appear to have been reported for chipping fracture, we may expect similar falloffs in critical bite force with cycling as in Fig. 7, i.e. a reduction of two or three over an equivalent loading year. Data for the critical bite force to induce chipping fracture from a hard, sharp contact in single-cycle loading are shown for selected dental ceramics in Fig. 8.^{110, 121} For small edge distances, critical force levels can be much lower than those for axial fracture modes, especially in porcelains, accounting for the prevalence of chipping in veneered restorations.

3. Discussion

The fracture mechanics approach advocated here offers a uniquely powerful base for evaluating fatigue properties of dental ceramics, with clear physical insight into a multitude of fracture modes. These competing modes have gone largely unconsidered in the dental community. The dental materials literature is replete with strength data from standardized single-cycle flexure tests on bar or disk specimens. Strength tests can be differently done, and some of the protocols may resemble the clinical situation more closely than others, but this does not mean that any one method is better—all can provide useful means of ranking materials. However, such tests do not come close to representing the long-term behavior of real prostheses. Single-cycle strength tests provide information only at the left axis of S - n diagrams such as Fig. 5, and therefore exclude information on those more deleterious fracture modes governed by mechanical degradation at later stages of cycling. Even S - n diagrams are limited in their usefulness, especially in crown configurations where the tensile stress states consist of a complex mixture of contact, flexural and membrane components and are sufficiently inhomogeneous as to cause newly initiated cracks to undergo the stages of arrest and stable propagation evident in Fig. 6. Simulated 'crunch-the-crown' tests with hard indenters, analogous to those represented in Figs. 3 and 4, take us a step closer to real restoration geometries, although proper caution needs to be exercised in linking laboratory observations to clinical situations²⁴. The widely popular techniques of finite element

modeling can usefully map out such complex stress states, but are inadequate to account for the stable fracture phase without the laborious incorporation of crack extension subroutines into the code.^{113, 122, 123} Nor can fractographic studies, so useful in identifying fracture origins, reveal much about the complex route from crack initiation to ultimate failure. Ideally, evaluations of lifetimes ultimately rest with tests on anatomically-correct specimens under conditions that replicate actual oral function, such as those in mouth-motion simulators,^{81, 82, 95, 124, 125} but these offer limited insight into the roles of the many controlling fatigue variables.

Lifetime fracture maps of the kind shown in Fig. 7, as well as delineating the regions of dominance for different fracture modes, provide guidelines for designing dental ceramic systems. In porcelain-veneered structures with lithium disilicate (Fig. 7a) and zirconia (Fig. 7b) cores, occlusal surface cracks (particularly inner cones) are dominant over the cycle range. In lithium disilicate monolith (Fig. 7c) and zirconia monoliths (Fig. 7d), radial (or margin) fracture dominates. Generally, zirconia-based monolithic or veneered structures are more damage resistant than are glass-ceramic-based, reflecting a higher toughness for the former material. Veneered structures have inferior lifetime characteristics relative to monoliths, partly because the weak porcelain is more susceptible to surface cracking and partly because the cracks have a smaller thickness to traverse to an interface. An important requirement in design is to maintain the lifetime trendlines above the range of natural bite forces, with maxima estimated variously between 100 N and 600 N.^{95, 126, 127} The veneered structures in Figs. 7c and 7d come close to violating this requirement, especially if porcelain chipping is factored in, indicative of an inherently vulnerable system.

Given our emphasis on mechanical fatigue in the long-term response of dental ceramics, some comment on the physical nature of the responsible mechanisms is called for. Mechanical degradation can manifest itself in periodic flexure testing, as in the S - n data in Fig. 5. For surfaces subject to point-contact damage, strength loss is due to degradation from internal friction followed by microcracking at weak interfaces within a near-surface damage zone.^{103, 128, 129} More pronounced mechanical fatigue occurs once the cracks enter the stage of stable propagation, e.g.

in the $c-n$ crack growth data for inner or partial cone and median cracks in Fig. 6. The principal underlying mechanism is then hydraulic pumping of aqueous solution into the fissures,⁶⁸ a kind of 'fracking'. A simple diagnostic in traditional fatigue testing for distinguishing mechanical from chemical (SCG) processes is to compare $c-n$ data obtained in cyclic versus steady or monotonic loading over comparable test durations: in single-cycle loading, outer cone and radial crack data sets remain parallel to the SCG growth trendlines, while inner and partial cone cracks and (usually) median cracks do not appear at all.⁶⁸

The bioengineering approach to lifetime evaluations described herein provides a strong physical basis for designing next-generation materials for dental prostheses. The key is a sound understanding of the roles of material and geometrical variables in damage accumulation in repetitive loading. Changes in these variables are manifested as shifts of different segments in the trendlines of Fig. 7. Materials design involves balancing several factors, which have been documented in the literature: (i) material properties,^{3, 12, 61, 65, 69, 91, 118, 130-136} (ii) microstructure,^{1, 10, 137-140} (iii) residual stresses,^{13, 15, 16, 135, 141-145} (iv) monolithic versus veneered structures,^{18, 19, 21, 22, 146} (v) layer thickness,^{12, 65, 90, 135, 147}, (vi) tooth contact conditions,^{12, 65, 66, 94, 148} (vii) tooth size and shape,^{91, 149} (viii) dentine, enamel and adhesive modulus,^{86, 150-157} (ix) surface state.^{63, 69, 91}

4. Conclusions

- (i) Model layer structures loaded with spherical indenters enable identification of clinically relevant fracture modes in layered dental prosthetic structures. Some of these modes are not easily inferred from conventional *post-mortem* examinations of failed parts.
- (ii) Ceramics are susceptible to loss of load-bearing capacity in cyclic loading, i.e. fatigue, amounting to declines in strength or critical bite force amounting to a factor of 2 or more over an equivalent one-year biting history.
- (iii) Part of fatigue is due to well-documented chemically-assisted slow crack growth (SCG), but more deleterious is degradation by mechanical processes such as hydraulic pumping and internal friction at microcrack walls. Some fractures, most notably inner cone cracks, do not appear at all

in static or monotonic loading.

(iv) Strength tests in cyclic flexure provide information on the stresses needed to initiate cracks, but are restrictive in information relating subsequent stable crack growth to ultimate failure.

(v) *In situ* fracture tests on transparent layer structures, coupled with rigorous fracture mechanics analysis of crack extension from initiation through stable growth to failure, facilitate construction of lifetime damage maps for common prosthetic material combinations.

(vii) Monolith structures are more resilient than their veneered counterparts. Zirconia is the most fatigue-resistant of the current dental ceramics.

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Figure Captions

1. Schematic diagram depicting various fracture modes in (a) crown and (b) FDP all-ceramic structures: axisymmetric outer (O) and inner (I) cone cracks, and median (M) cracks; partial cone (P) cracks; edge chipping cracks (C); radial (R) cracks at cementation surfaces; flexure (F) cracks at connectors. Linear-trace cracks (O, I, P, C, F) extend out of the plane of diagram, shaded (R, M) cracks extend within the plane of diagram.
2. Fractures in dental prostheses. Figures (a) through (c) are clinical failures: (a) porcelain-veneered zirconia molar crown, showing crack originating from wear facet on occlusal surface; (b) longitudinal fracture from margin of Dicor crown (courtesy K. Malament); (c) connector fracture between 2 pontics of a 4-unit porcelain-veneered FDP. Figures (d) through (f) are laboratory failures of porcelain-veneered zirconia prostheses: (d) side view of veneer chipping in off-axis loaded crown, single-cycle loading with sharp point (Vickers indenter) at load 700 N; (e) post-testing section view of partial cone crack in crown veneer loaded centrally and tangentially, after 6×10^4 cycles with sphere indenter at 300 N (Courtesy P. Guess); (f) fracture at connector of 3-unit FDP, after 7.8×10^4 cycles with sphere indenter at 700 N (Courtesy C. Stappert).
3. Occlusal-surface damage in flat-layer porcelain specimens in contact with a hard sphere. (a) Post-indentation half-surface (top) and section view (bottom), monolith porcelain specimen (Vita Mark II, Vita Zahnfabrik) loaded axially at 5×10^4 cycles. Note deep-penetrating cone cracks from inner contact zone.¹⁵⁸ (b) Post-indentation half-surface (top) and section view (bottom), flat specimen porcelain-fused-to-zirconia (LAVA, 3M ESPE) loaded tangentially at 7.8×10^4 cycles. Sliding direction left to right.²⁰ Note partial cone cracks approaching porcelain/zirconia interface.
4. Longitudinal fracture modes in model transparent layer structures bonded to dentin-like polymeric bases. Cracks in (a) and (b) are visible as series of interference fringes from light source. All images except (d) are *in situ* views with load applied axially with a hard sphere. (a) Radial crack initiated in glass layer at cementation surface.⁸⁵ (b) Radial crack within sapphire core arrested at glass/sapphire interface.⁹⁰ (c) Radial cracks in glass shell backfilled with epoxy resin.⁹¹ (d) Similar glass shell to (c), but now with top surface

loaded off-axis with polymer disk to emulate food bolus.¹¹⁸ In all cases the cracks remain within a single brittle layer, increasing incrementally in size with occlusal load. Severe overload can lead to delamination or penetration into the sublayer.

5. Strength of dental zirconia ceramic (Prozr Y-TZP) plates bonded to a polycarbonate base in flexural loading, as a function of cycles to failure at frequency 10 Hz. Data shown for surfaces in polished, sandblasted and point-load-indented states. Linear trendlines are in accord with degradation from chemically-assisted slow crack growth in presence of water. Downward deviations from these linear trendlines indicate superposed degradation from mechanical fatigue. From.⁶⁷
6. Crack depth through layer section versus number of cycles for a monolith flat glass plate of thickness 1 mm bonded to a thicker polycarbonate base and loaded with an axial force 120 N at its top surface with a hard sphere of radius 1.6 mm at frequency 1 Hz (cf. Fig. 4a). (a) Cone and median cracks initiated at glass top surface. (b) Radial cracks initiated at glass bottom surface. Solid inclined line indicates expected growth rate due solely to SCG, for token crack initiated to depth 0.5 mm on first cycle. Vertical dashed lines indicate abrupt initiation stages.⁶⁶
7. Bite force to penetrate layer thickness versus number of cycles for flat layer structures of net thickness 1.5 mm bonded to dentin and loaded axially with a porcelain sphere (or opposing tooth) of radius 5 mm. Trendlines evaluated from fracture mechanics lifetime equations, in conjunction with data extrapolation from Fig. 6.¹¹⁸ Estimates for (a) lithium disilicate and (b) zirconia cores veneered with 1 mm porcelain, and (c) lithium disilicate and (d) zirconia monoliths. Trendlines can shift according to key material and geometrical conditions.
8. Critical bite forces to form chipping fractures from sharp contacts in dental ceramic materials, as function of distance from side wall. Lines from chipping equations from fits to data on flat monolith and anatomically-correct crown specimens.^{110, 121} The relative locations of the lines highlight the toughness of zirconia ceramics and weakness of porcelains, with alumina and lithium disilicate intermediate.

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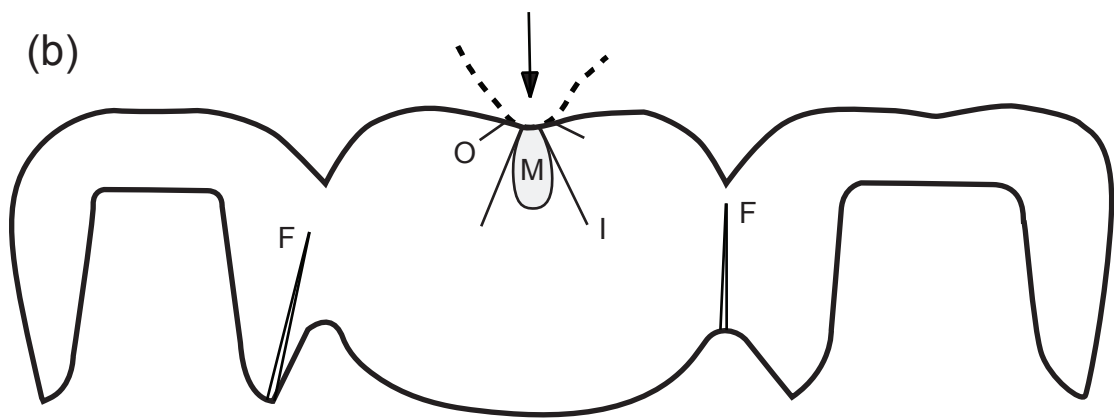
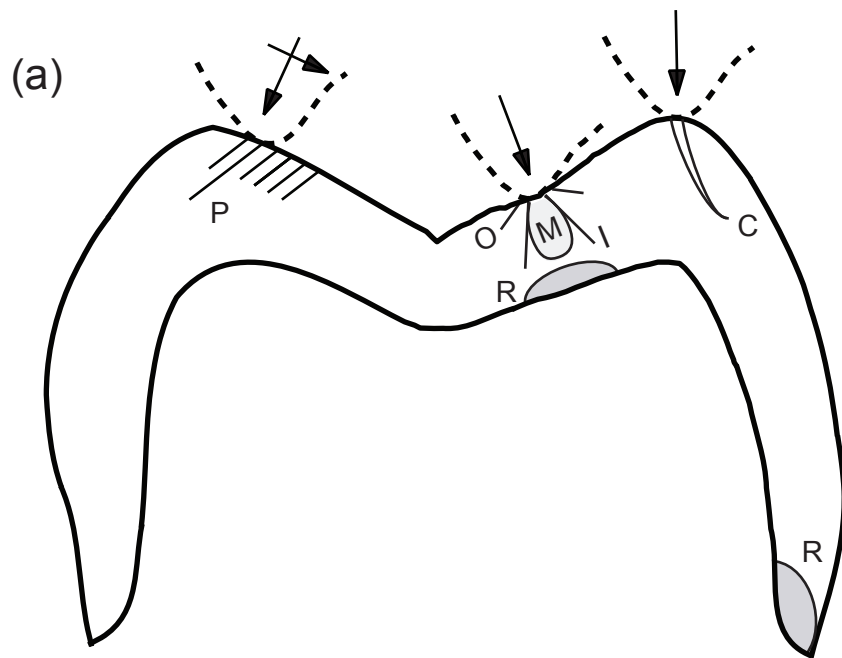
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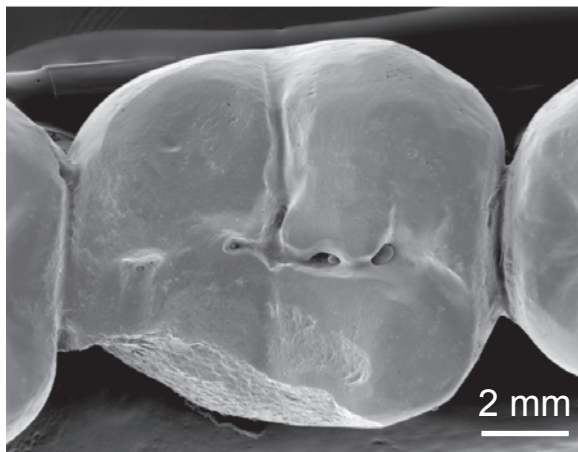
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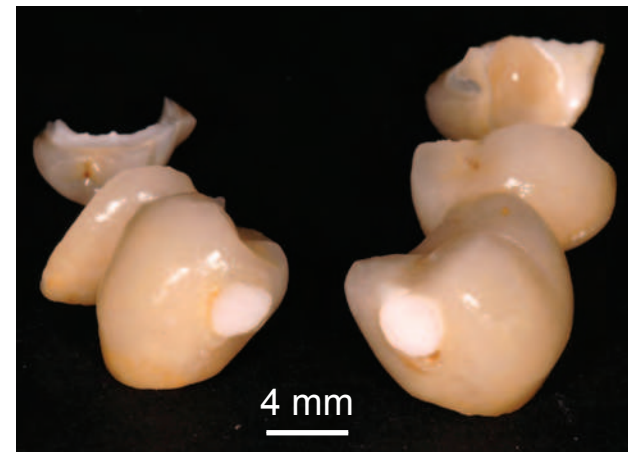
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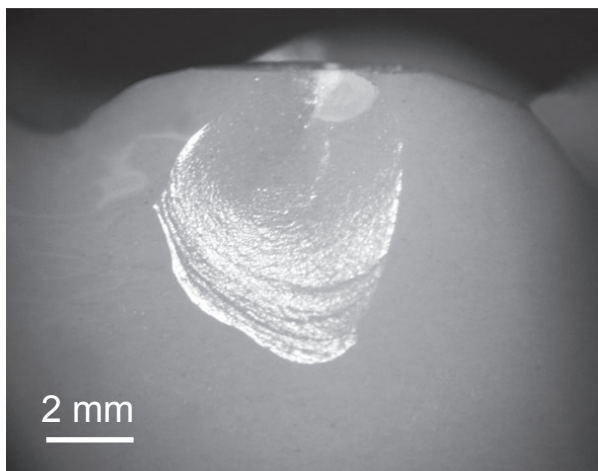
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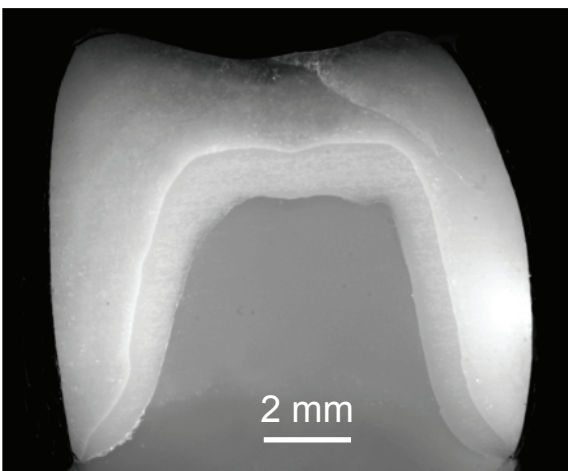
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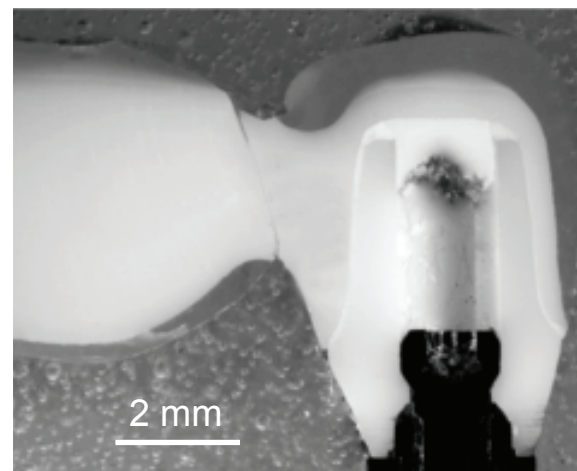


Fig. 2

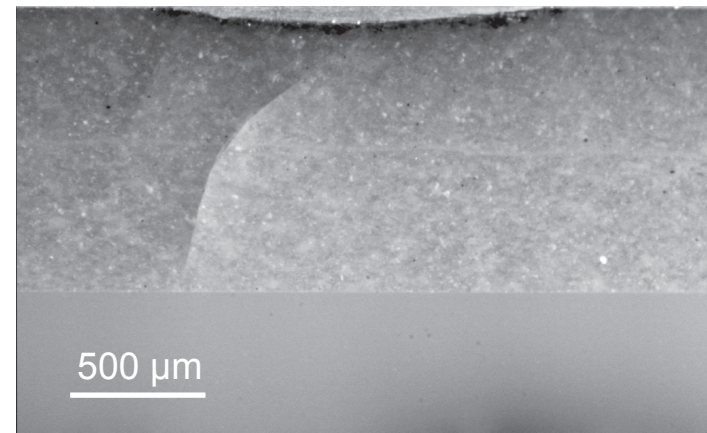
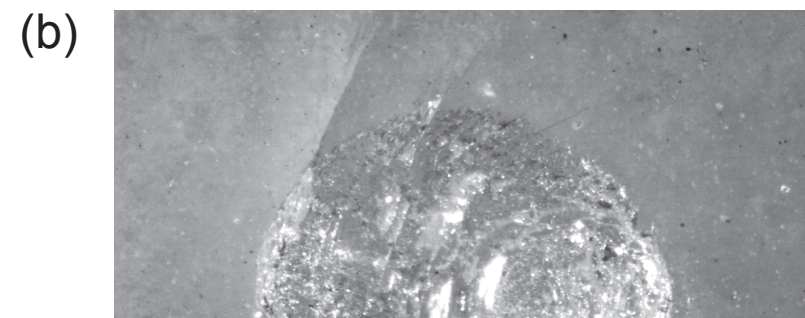
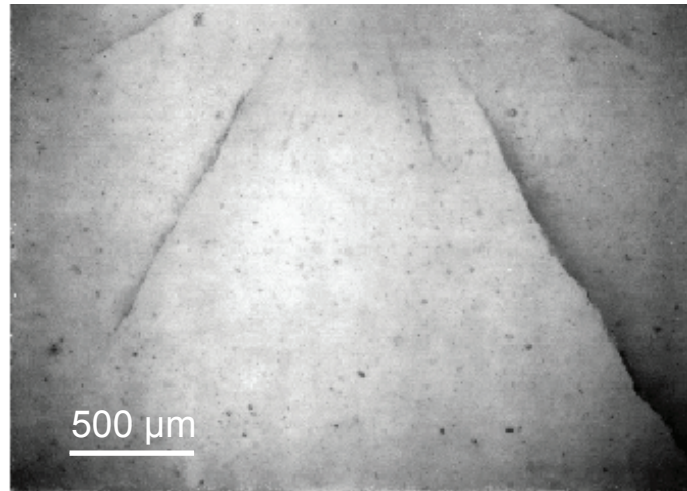
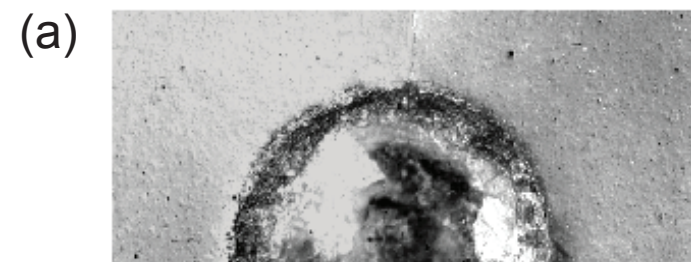


Fig. 3

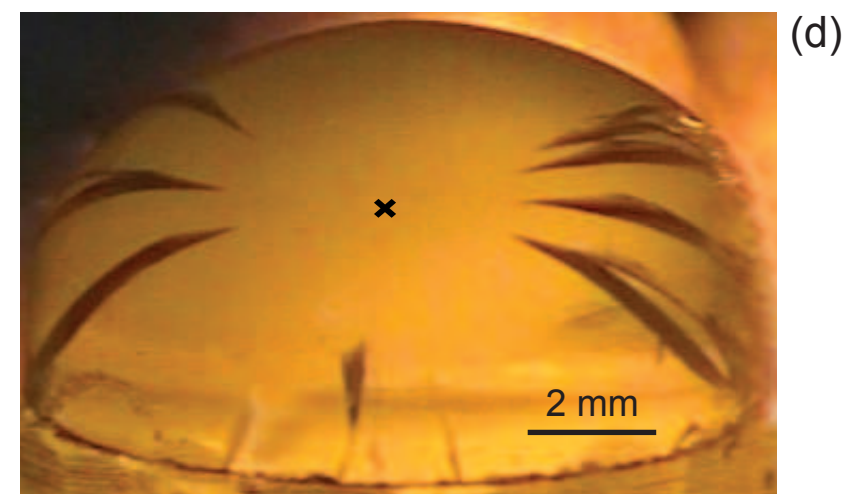
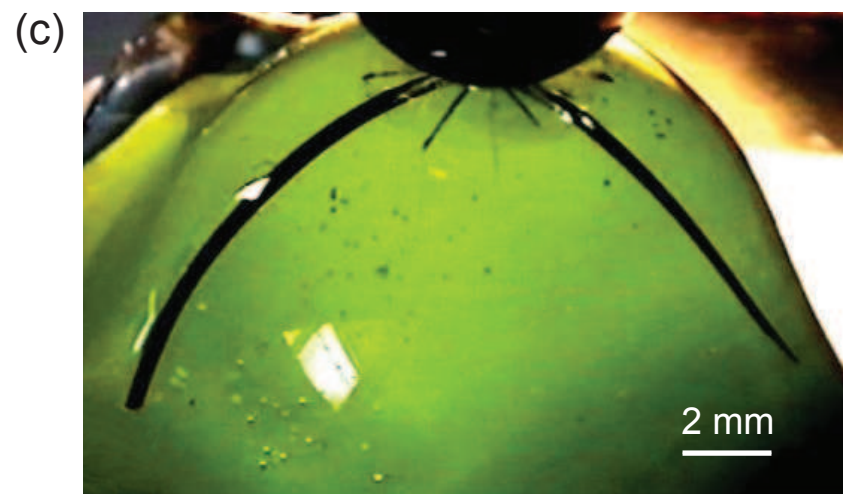
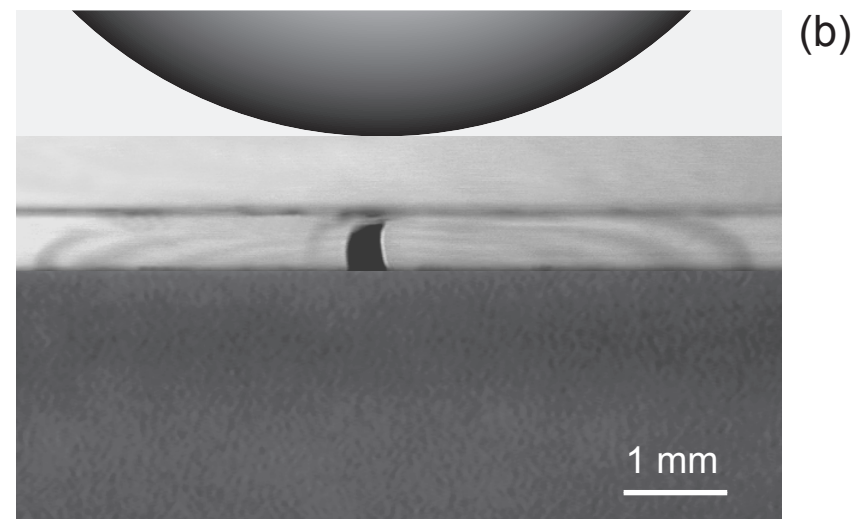
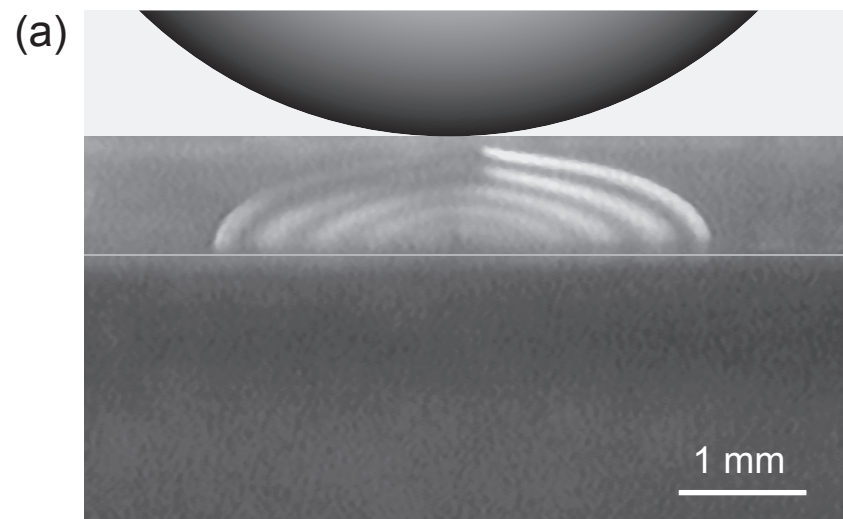


Fig. 4

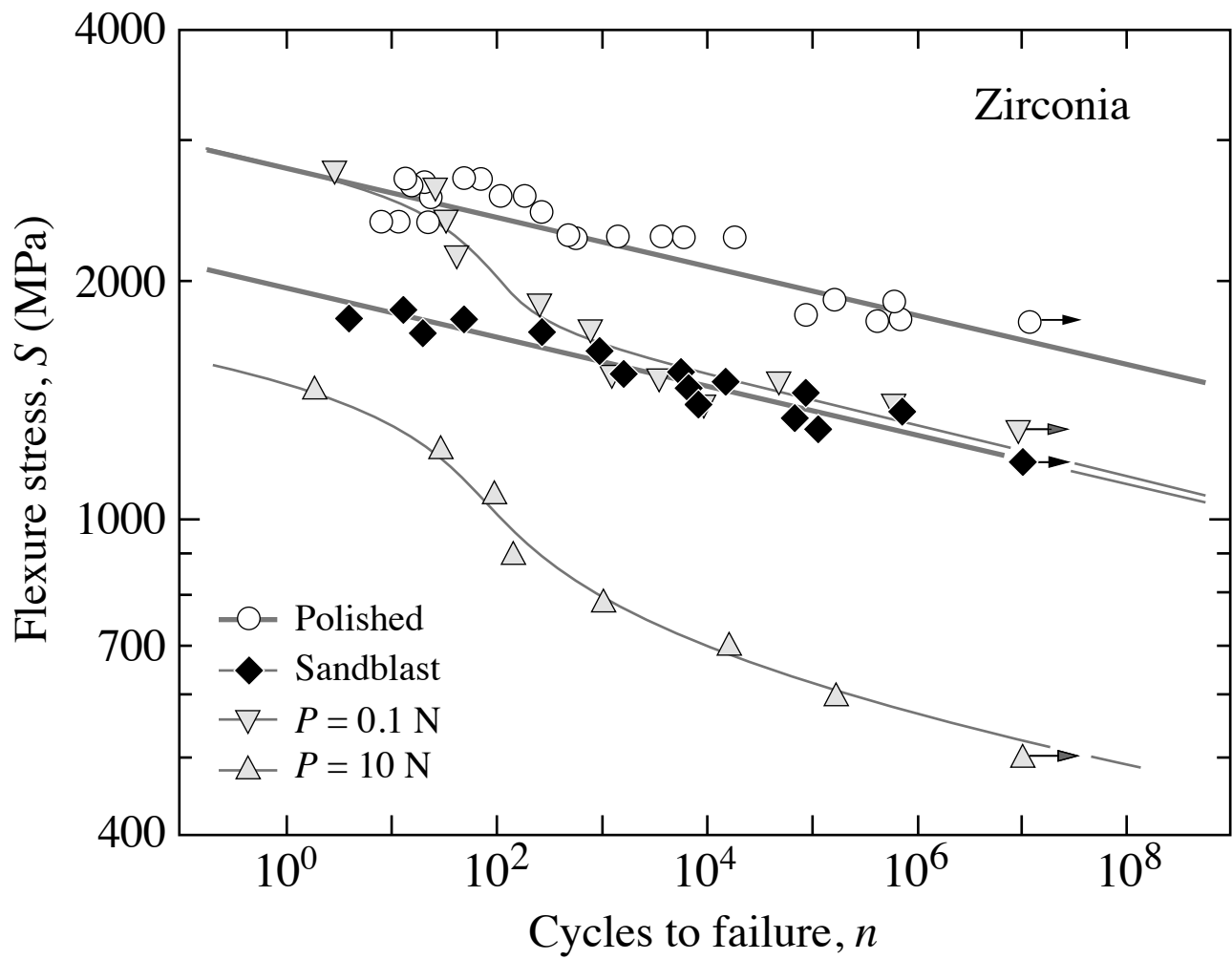


Fig. 5

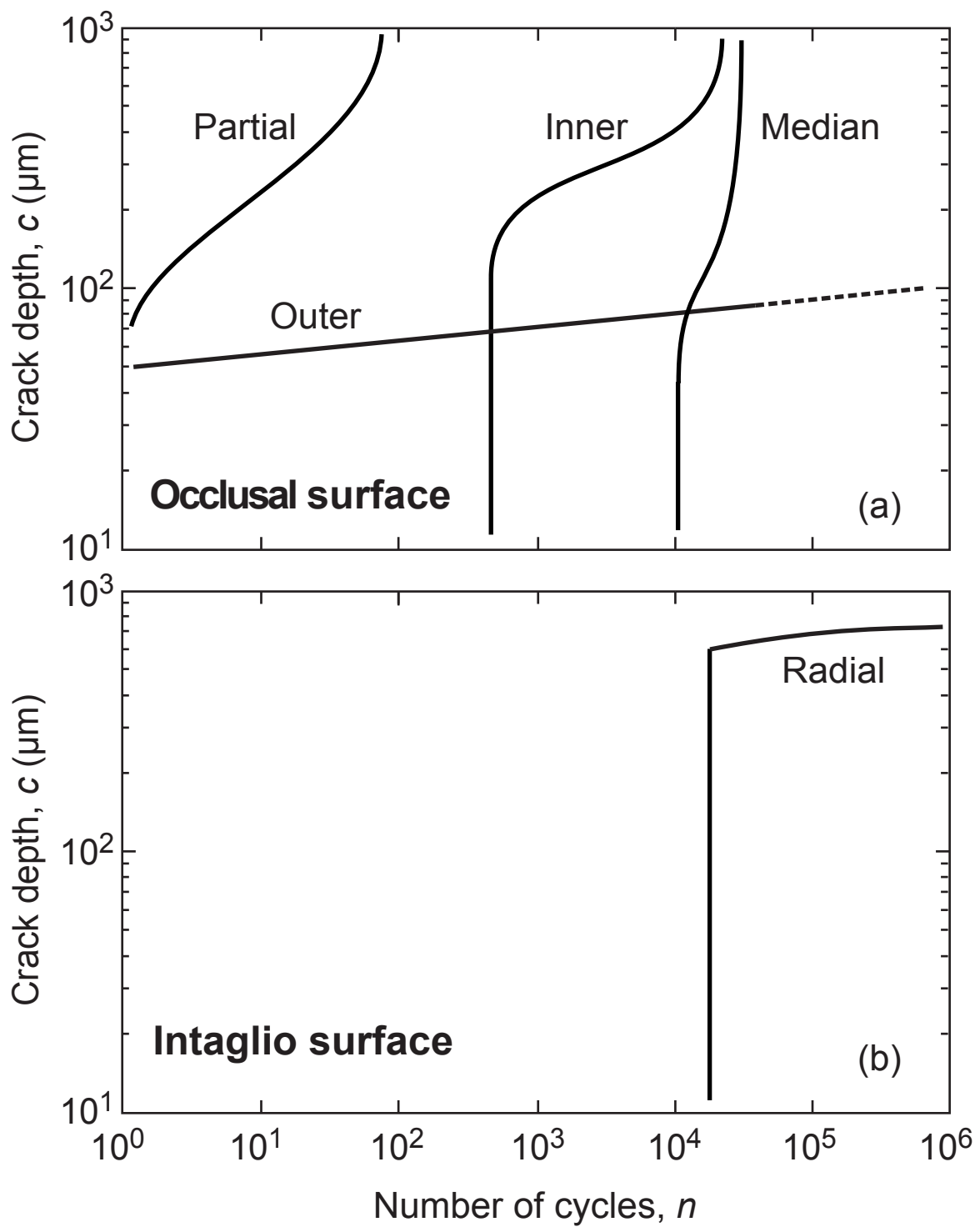


Fig. 6

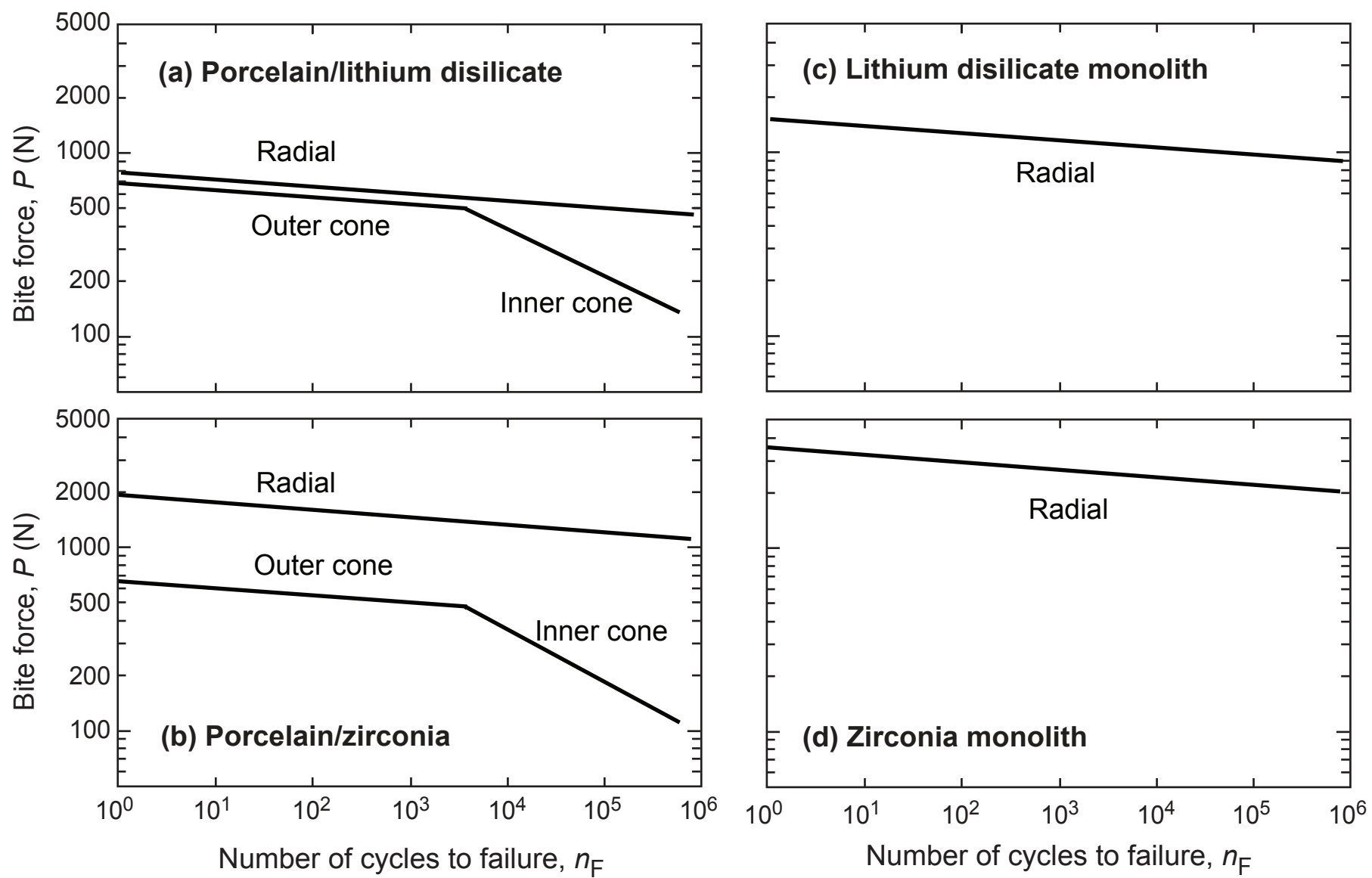


Fig. 7

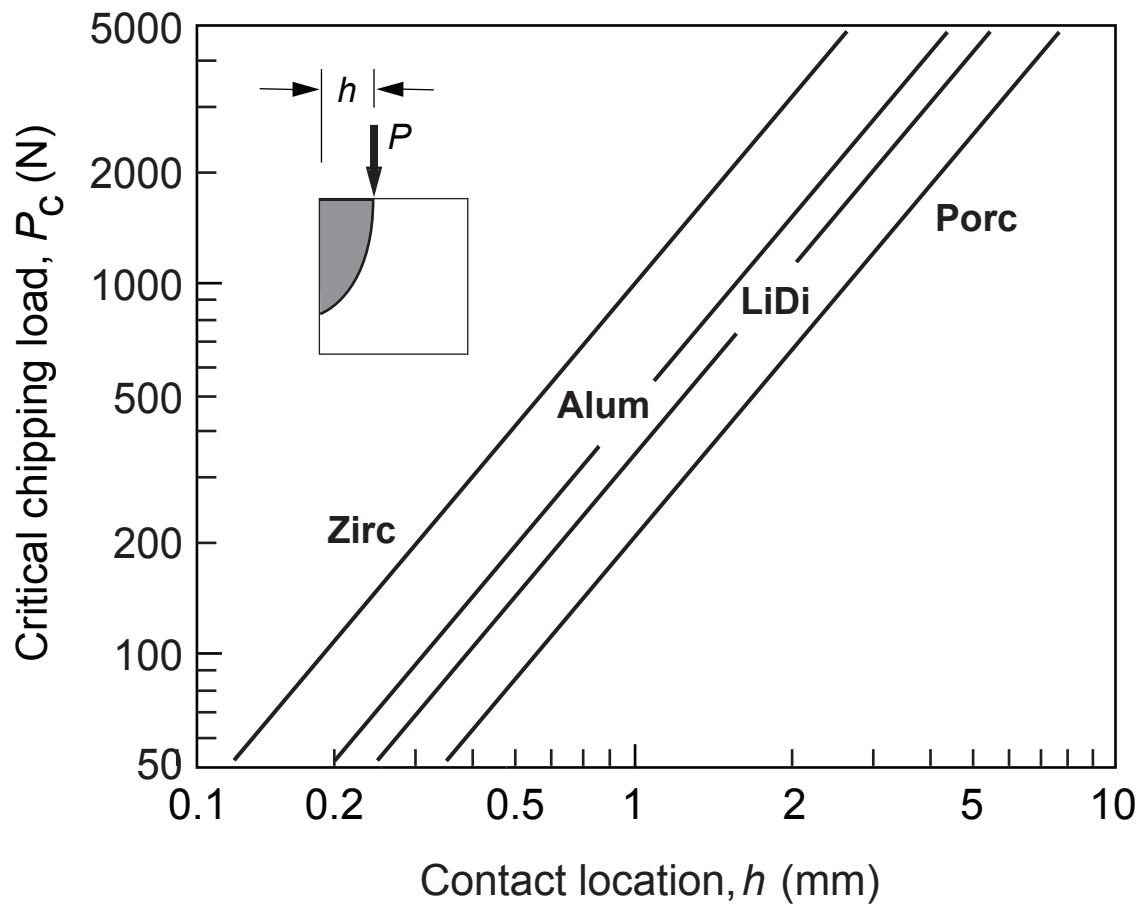


Fig. 8